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Control of lateral balance in walking

Experimental findings in normal subjects and above-knee amputees

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Abstract

In walking the human body is never in balance. Most of the time the trunk is supported by one leg and the centre of mass (CoM) ‘falls’ to the contralateral side. In dynamical situations the velocity of the CoM should be acknowledged as well in the ‘extrapolated centre of mass’ (XcoM). Centre of pressure (CoP) position was recorded by a treadmill with built-in force transducers. Lateral CoM and XcoM position were computed by filtering the CoP data. Subjects were six above-knee amputees and six matched healthy controls. They walked at approximately 0.75, 1, and 1.25 m/s for 2 min.

Amputees showed asymmetric gait with shorter stance (60%) at the prosthetic side versus 68% at the non-prosthetic side and a wider stride (13 ± 4 cm, mean \pm S.D.) compared to controls (9 ± 3 cm). At foot placement CoP was just lateral to the XcoM. The margin between average CoP and XcoM at foot contact was only 1.6 ± 0.7 cm in controls, 2.7 ± 0.5 cm in amputees at the prosthetic side and 1.9 ± 0.6 cm at the non-prosthetic side. Next to this ‘stepping strategy’, CoP position was corrected after initial contact by modulating the lateral foot roll-off (‘lateral ankle strategy’) in non-prosthetic legs up to about 2 cm.

A simple mechanical model, the inverted pendulum model, can explain that: (1) a less precise foot placement (greater CoP–XcoM margin) results in a wider stride, (2) this effect can be reduced by walking with a higher cadence, and (3) a greater margin at one side, as with a leg prosthesis, should be compensated by a shorter stance duration at the same side to achieve a straight path. This suggests that not in all cases symmetric gait should be an aim of rehabilitation.

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Keywords: Inverted pendulum model; Stepping strategy; Ankle strategy; Equilibrium; Gait

1. Introduction

Two-legged walking poses a difficult balance control problem. Most of the time the trunk is supported by one leg only and the whole body center of mass (CoM) is never above the base of support. This essentially unstable system can only be stabilized by active control. Previous studies on models comprising the complete three-dimensional mechanics of walking [1–4] have shown that forward and lateral movements in walking are to a large degree independent, with the significant exception that stride time is controlled by the

forward movement. With stride time fixed, lateral motion is unstable, unless foot placement is controlled. The models involved are variants of the ‘inverted pendulum’ model: the body is modelled as a single mass, concentrated in the CoM, balancing on a rod the lower end of which is put on the ground at the ‘center of pressure’ (CoP), somewhere under the foot. The parameter of the inverted pendulum model is its eigen frequency $\omega_0 = \sqrt{g/l}$. The left and right feet are positioned alternately some distance apart, while the CoM is in between. When standing on the left foot the CoM falls to the right and vice versa, but the fall is always reversed timely [5]. According to the above, balance in walking is said to be maintained by a ‘stepping strategy’ [6,7].

The aims of the present paper are to verify this assumption experimentally and to investigate which control

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Nomenclature

b_{\min}	minimal distance between CoP and XcoM in a step, usually at foot contact (cm)
CoM	projection of the center of mass on the ground, symbol $z(t)$
CoP	center of pressure, effective position of the point of attack of the ground reaction force vector, symbol $u(t)$
$\cosh(x)$	hyperbolic cosine, $\cosh(x) = (e^x + e^{-x})/2$
g	acceleration of gravity = 9.81 m s^{-2}
h	effective height of the body CoM above the floor = $1.34l$ (m)
l	leg length = height of greater trochanter above the floor (m)
$\sinh(x)$	hyperbolic sine, $\sinh(x) = (e^x - e^{-x})/2$
T	step time (s)
$u(t)$	lateral position of CoP (m)
u_L, u_R	CoP position of left and right foot, respectively, assumed constant during the step
u_{ms}	measured CoP position, averaged over a step
v_0	CoM velocity = \dot{z} at foot contact, (m s^{-1})
XcoM	extrapolated center of mass, symbol $\zeta(t)$, with $\zeta(t) = z(t) + (1/\omega_0) \cdot (dz/dt)$
$z(t)$	lateral position of CoM (m)
\dot{z}	lateral velocity of CoM (m s^{-1})
\ddot{z}	lateral acceleration of CoM (m s^{-2})
<i>Greek symbols</i>	
$\zeta(t)$	position of XcoM (m)
ω_0	pendulum eigen (angular) frequency $\sqrt{g/h}$ (rad s^{-1})

law is used in this foot placement: where exactly is the foot to be placed to achieve a stable gait? Another question is to see if any other balance strategies are used in addition, e.g. ankle strategy or hip strategy. The subjects were a group of above-knee amputees with leg prosthesis and a matched normal control group. Prosthesis walkers have been chosen because they have definite balance problems, probably associated with a lack of control of the ankle moment. A Stroop test was included in the experimental protocol, to see if the subjects needed cognitive attention during walking.

The condition for standing stability is usually formulated as: the vertical projection of the CoM on the ground should be within the base of support. The base of support is loosely defined as the area between the feet. We have recently shown that this condition should be reformulated to relate not to the CoM position alone, but to the position of the ‘extrapolated center of mass’ (XcoM) which is equal to CoM position plus CoM velocity/ ω_0 [8]. According to the inverted pendulum model, CoM acceleration is proportional to the distance between CoP and CoM-projection. If the CoM has too large a velocity towards the CoP, however, the acceleration is

insufficient to reverse the direction of CoM movement. For walking no base of support can be defined, but the actual CoP positions of both feet can be measured. When lateral CoM position is denoted by z and CoP position by u , it should thus hold that:

$$u_L < z + \frac{\dot{z}}{\omega_0} < u_R \quad (1)$$

$$\text{CoP}_{\text{Left}} < \text{XcoM} < \text{CoP}_{\text{Right}}$$

in which the left hand inequality holds when the left foot is on the ground and the right hand one for the right foot.

2. Methods

2.1. Treadmill

Recordings were made by means of an instrumented treadmill [9]. The treadmill walking surface was divided into a left and a right half, each provided with four transducers for measuring the vertical ground reaction force. From the distribution of the forces the CoP can be calculated. It was verified that this procedure is accurate within 0.6 cm. Data acquisition was done by a 12-bits A/D card at 50 Hz under control of a LabView program, data processing was done by a custom program written in MatLab.

The projection of the center of mass (CoM) at ground level was computed from the CoP data by low-pass filtering [10–13]. This method is based on the inverted pendulum model of human balance [14,15] and assumes that angular accelerations of the trunk can be neglected. The only parameter of this method is $\omega_0 = \sqrt{g/h}$. For the equivalent pendulum length h in lateral motion a value of 1.34 times trochanteric height l was taken [16]. It was previously shown by comparison with kinematical methods that this approximation is valid in human walking at the usual speeds. Velocity of the CoM was obtained from CoM position by numerical differentiation with a low-pass cut-off at 4 Hz. Temporal data, heel contact, toe-off, etc. were determined on the basis of the forward CoP velocity. Mean values u_{ms} for CoP position u over a step were calculated as averages over the period of single stance = contralateral swing. All experiments were also recorded on video in a view from behind.

In the present paper only results on lateral movement will be presented. As a consequence, it is not necessary to carefully discriminate between the actual CoM position, somewhat above-hip level, and its projection on the ground. In accordance with the ISB recommendations [17] lateral position is presented as the z -coordinate, positive to the right.

2.2. Subjects, procedure

The subject group consisted of six experienced (6–40 yr) above-knee amputee walkers, four men, and two women,

Table 1
Subject data

Amputees	Sex	Age (yr)	Since (yr)	Side	Mass (kg)	Stature (m)	Leg length (m)	Controls	Sex	Age (yr)	Mass (kg)	Stature (m)	Leg length (m)
1	M	32	15	Left	55	1.65	0.83	7	M	26	67	1.78	0.83
2	M	43	6	Left	60	1.73	0.86	8	M	50	70	1.67	0.87
3	M	41	40	Right	61	1.82	1.00	9	M	53	92	1.84	1.00
4	M	43	25	Left	111	1.86	0.95	10	M	55	94	1.92	1.00
5	F	34	25	Left	60	1.67	0.96	11	F	55	78	1.78	0.92
6	F	50	36	Left	69	1.72	0.86 ^a	12	F	21	61	1.62	0.82

Personal data of amputee subjects and matched controls. Leg length was measured from greater trochanter to floor.

^a Subject 6 wore shoes with a 6 cm heel.

and six control subjects, matched by leg length, mass, and sex (see Table 1).

Subjects walked at three speeds for periods of 2 min, with 5 min of rest in between. Walking speeds were selected as 0.75, 1.00, and 1.25 m s⁻¹ for a leg length of 1.00 m. For subjects with other leg lengths l (in meters), speed was multiplied with \sqrt{l} . In this way normalized speed equalled 0.24, 0.32, and 0.40 for all subjects [18]. Amputee subject 2 was not able to walk at the ‘fast’ speed. After the first series of three speeds, the procedure was repeated, but now the subjects had to perform a Stroop test while walking. For this test words like “red”, “blue”, “green”, projected in non-matching colors, were presented on a computer display 1.50 m in front of them, at a pace of one word per two seconds. Subjects were then asked the color of the text. Subjects were asked not to use the side bars of the treadmill, but the amputee subjects could not fully comply with this request. They were instructed to hold it as lightly as possible and not to lean on it. All subjects were secured against falling by a safety harness connected to a rail at the ceiling. The experimental protocol was approved by the local Medical Ethics Committee and the subjects gave their written consent.

3. Results

3.1. Temporal factors

Both amputees and controls showed a decrease of stride time (i.e. an increase of cadence) with speed, but the decrease was less for amputees (Table 2). At the ‘normal’ speed of 1 m/s stride time was longer in amputees. Gait was markedly asymmetric in the amputee group: stance was shorter for the prosthetic leg, 60.4% of stride (range 57.4–64.6%), versus 68% for the non-prosthetic leg (range 65.9–70.1%), while in the control group both were on average 64%. Only one amputee subject showed a symmetry comparable to the control group: 64.6 and 65.9% for prosthetic and non-prosthetic leg, respectively. Double contact times showed no differences between amputees and controls and between legs (Table 2).

3.2. Spatial data

In Fig. 1 examples of recorded CoP registrations are shown. In left and right single stance CoP position only changes little, while it traverses quickly to the contralateral

Table 2
Results

	Amputees		Controls	
	Prosthetic leg	Normal leg	Left	Right
Stride time (s)				
Slow	1.51 (0.13)		1.51 (0.17) s*	
Normal	1.35 (0.13)		1.31 (0.11)	
Fast	1.29 (0.10)		1.19 (0.08) s*	
Stroop	1.35 (0.13)		1.34 (0.09)	
Stance (percent of stride)				
Slow	59.4 (1.1) a** p**	67.4 (1.7) a*	63.1 (1.6)	65.3 (0.9)
Normal	60.4 (3.0) a* p**	68.0 (1.6) a**	64.1 (0.9)	64.4 (0.9)
Fast	58.5 (2.9) a** p**	67.5 (1.5) a**	64.1 (0.6)	64.3 (0.9)
Stroop	60.3 (3.0) a** p**	67.8 (2.1) a**	64.3 (0.7)	64.8 (1.2)
Double contact (percent of stride)				
Slow	13.6 (1.1)	14.1 (2.1)	14.3 (1.1)	14.2 (1.0)
Normal	14.4 (2.3)	14.2 (1.5)	13.9 (0.9)	14.8 (0.8)
Fast	13.4 (2.6)	12.8 (1.3)	13.6 (0.7)	14.9 (0.5)
Stroop	14.7 (2.7)	13.6 (1.7)	14.1 (0.9)	15.1 (0.7)
Stride width (cm)				
Slow	12.3 (3.0) a*		8.2 (3.5)	
Normal	12.9 (4.0) a*		8.6 (3.3)	
Fast	14.7 (4.8) a*		8.8 (2.5)	
Stroop	14.4 (4.6) a*		8.8 (2.4)	
b_{\min} Mean (cm)				
Slow	2.42 (0.36) a** p**	1.62 (0.42)	1.40 (0.72)	1.35 (0.74)
Normal	2.74 (0.54) a** p*	1.90 (0.62)	1.61 (0.71)	1.67 (0.70)
Fast	3.25 (0.88) a** p**	2.20 (0.92)	1.81 (0.61)	1.86 (0.57)
Stroop	2.99 (0.46) a** p*	2.11 (0.69)	1.65 (0.60)	1.66 (0.57)
b_{\min} S.D. (cm)				
Slow	0.403 a* p**	0.289	0.325	0.319
Normal	0.384 p**	0.278 a*	0.378	0.362
Fast	0.477 a* p**	0.320	0.403	0.363
Stroop	0.400 p**	0.301	0.338	0.302 s*

Results on temporal factors, stride width w , and CoP–XcoM distance b_{\min} , mean (S.D.); a*, significant difference between amputees and controls with $p < 5\%$; p*, significant difference between prosthetic leg and normal leg with $p < 5\%$; s*, significant difference between this speed or Stroop test and normal speed, no Stroop test, with $p < 5\%$; a**, etc. $p < 1\%$. Stride time has been divided by \sqrt{l} , i.e. normalized for a leg length of 1.0 m (see Section 2).

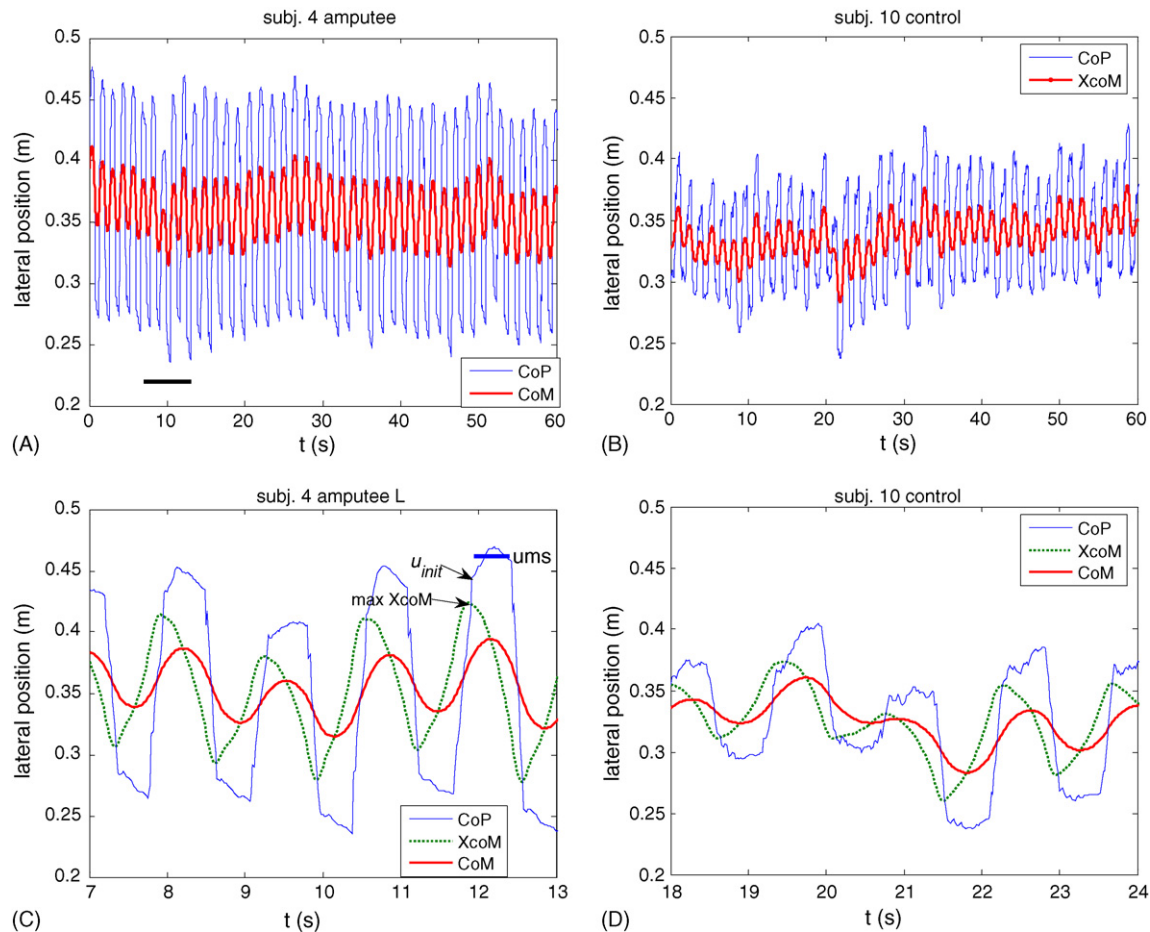


Fig. 1. (A and B) Recording of center of pressure (CoP, thin lines) and center of mass (CoM, thick lines) during the first 60 s of recording in a subject with a left side above-knee prosthesis (A) and his matched control (B), respectively. The same subjects have been presented in Figs. 3, 5 and 6. (C and D) CoP and CoM as in (A and B), but now on a 5-s timescale. Added is the extrapolated center of mass (XcoM, thick dotted lines).

foot in the double stance period. It is clearly seen that the presented amputee subject (Fig. 1A and C) showed a wider stride than the matched control (Fig. 1B and D). Although not equally extreme, this was the case in all subject pairs (Fig. 2) and it also turned out from the average (Table 2). Nevertheless, individual stride widths could differ over a factor of two in the control group as well (Fig. 2). Remarkably, the female subjects in both the amputee groups (5 and 6) as in the control groups (11 and 12) showed the smallest stride width. Foot (CoP) placement showed at times quite sudden variations, up to 5 cm, which were corrected in a few subsequent steps. As a consequence, stride width can be very different in consecutive strides. Average left and right stride width are the same for a straight path, of course. The CoM follows the CoP excursions in phase, but with a lower amplitude, about 25% of CoP at 1 m/s. CoM trajectory remained within a range of about 10 cm around the middle of the treadmill belt.

The time course of CoP, CoM, and XcoM is presented in more detail in Fig. 1C and D. Lateral CoM position shows a sinus-like smooth pattern, in phase with the alternating left-right square-wave pattern of the CoP. The XcoM trajectory is

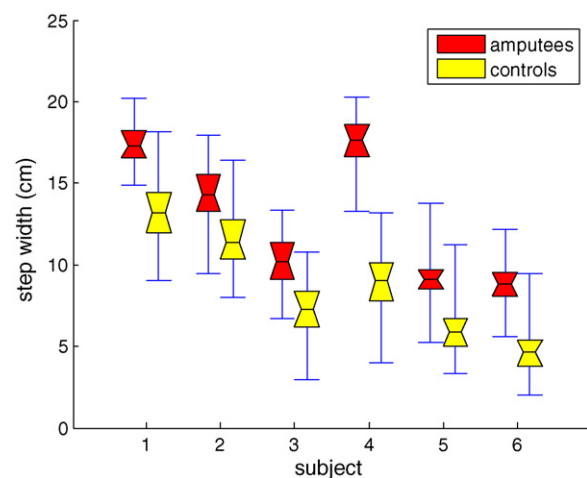


Fig. 2. Boxplot of step widths in amputees (subjects 1–6) and matched controls (subjects 7–12) drawn next to 1–6. Boxes give, from bottom to top, minimum, 25th percentile, median, 75th percentile, maximum.

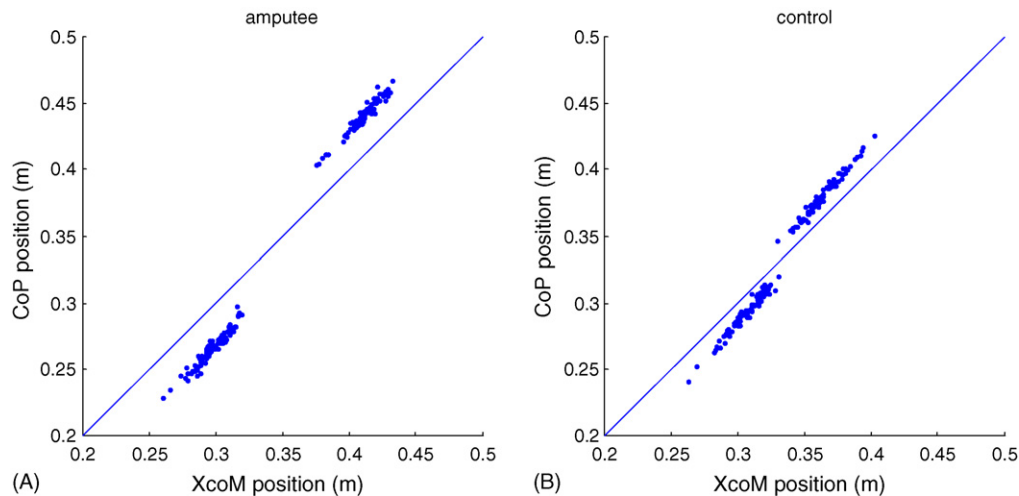


Fig. 3. CoP position μ_{ms} , averaged over each step, against XCoM position at the instant of foot contact. It is seen that CoP position is always slightly lateral to the XCoM, both for the amputee subject (A) as for the control (B), but that this distance (b_{min}) is higher in the amputee.

less smooth and has extremes around the times of foot contact. At every new step the CoP is placed only a small distance lateral to the current XCoM position. After foot placement the XCoM turns sharply towards the contralateral side. In Fig. 3 average CoP positions have been plotted against XCoM position at the time of foot contact for the two recordings of Fig. 1. It is seen that in amputees (Fig. 3A) CoP–XCoM distance is greater than in controls (Fig. 3B). Fig. 4 shows a box plot for this distance, b_{min} , for all subjects and Table 2 gives the averages. In amputees b_{min} for the prosthetic leg was always larger than for the non-prosthetic leg and larger than the values for the control subjects. In the control group b_{min} was closely equal for both legs, but with considerable interindividual differences. There was no significant difference in b_{min} between non-prosthetic legs of amputees and controls. Neither b_{min} nor stride width

showed a significant effect of walking speed. The Stroop test did not give an effect either.

The CoP recording of Fig. 1C shows at the left (prosthetic) side a stereotypical pattern during stance, as could be expected from a prosthetic foot without ankle musculature. In contrast, at the right (non-prosthetic) side the CoP patterns during single stance were much more variable from step to step. This was even more evident in the control subject of Fig. 1D. If initially the CoP was placed too close to, or even within the XCoM (e.g. second right step in Fig. 1D) it moved quickly outward. If it was initially already at a sufficient distance (as in the third right step in Fig. 1D) CoP remained constant, or even moved inward in some subjects. To illustrate this effect Fig. 5A and B shows lateral CoP position as a function of time minus XCoM at foot contact for all (about 100) steps of the recordings of Fig. 1A and B. In the normal feet it is seen that CoP moved outward if the initial position was too close to the XCoM (solid lines) and inward in the opposite case (dotted lines). In the prosthetic foot (Fig. 5A, lower part), all traces were more or less parallel. In several steps the initial CoP–XCoM distance was negative, i.e. CoP was initially within the XCoM, but in all cases b_{min} , CoP–XCoM averaged over the step, was positive, even if it could amount at times to only a few millimetres (Fig. 4). Fig. 6 shows a scatter diagram of CoP motion, final minus initial CoP position, as a function of initial CoP–XCoM distance. Normal legs showed a negative correlation: if the initial CoP position was too close, CoP moved outward during stance, if the initial CoP position was too far from the XCoM, CoP moved inward. In the prosthetic leg this correlation was around zero. Data on the correlation coefficients of all subjects are in Table 3. A consequence is that the standard deviation in the average margin b , is considerably smaller than the S.D. of the initial value, at foot placement (see Table 4). This effect is most pronounced in the non-prosthetic leg of the amputees, even more than in many of the control subjects.

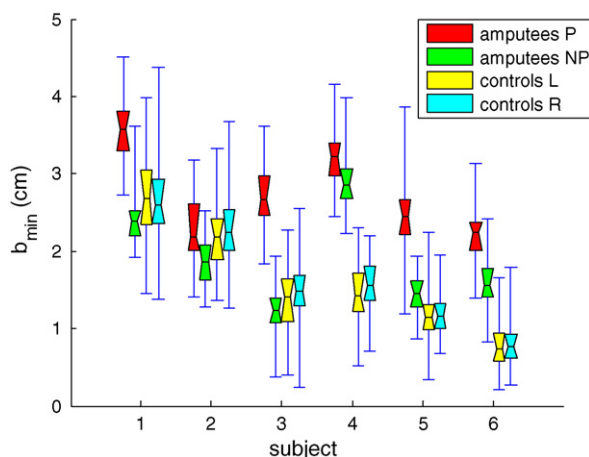


Fig. 4. Boxplot of minimum distance between CoP and XCoM at foot contact. Shown are: from left to right, amputee subject (1–6), prosthetic leg, amputee normal leg, control subject (7–12) left and right leg. On average in amputees b_{min} is larger in the prosthetic leg, compared to the contralateral leg, and to the legs of the controls. In the amputee normal leg it is in the range of the controls (see Table 2).

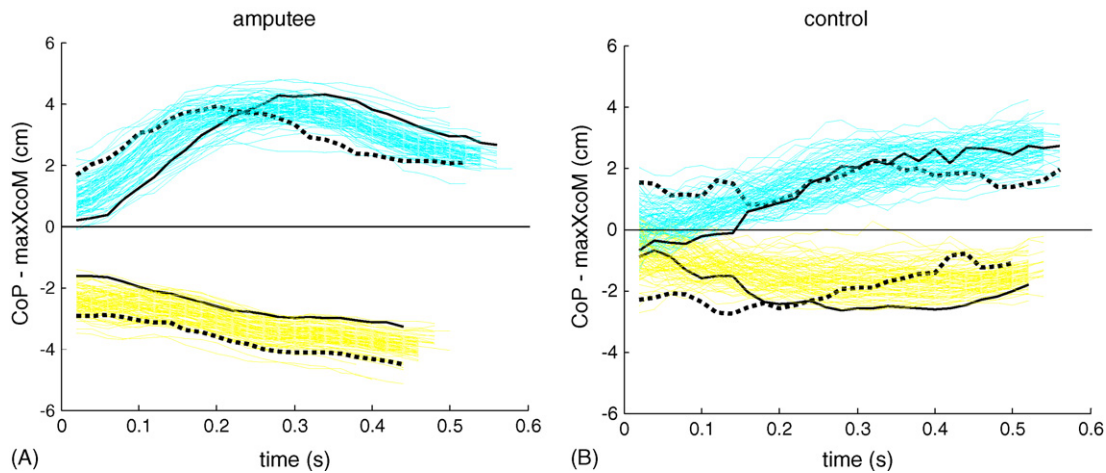


Fig. 5. Lateral CoP position during stance minus XcoP at the time of foot contact for (A) an amputee and (B) a control subject. Upper half: right foot, lower half, left foot. The average trajectory of the CoP is more or less stereotypical (band of lines), but a trajectory that starts (too) close to the XcoM (drawn thick line) moves more outward. If it has started too much outward, it ends more inward (dotted line). This effect, however, is not seen in the prosthetic leg (A), lower (left foot) trajectories.

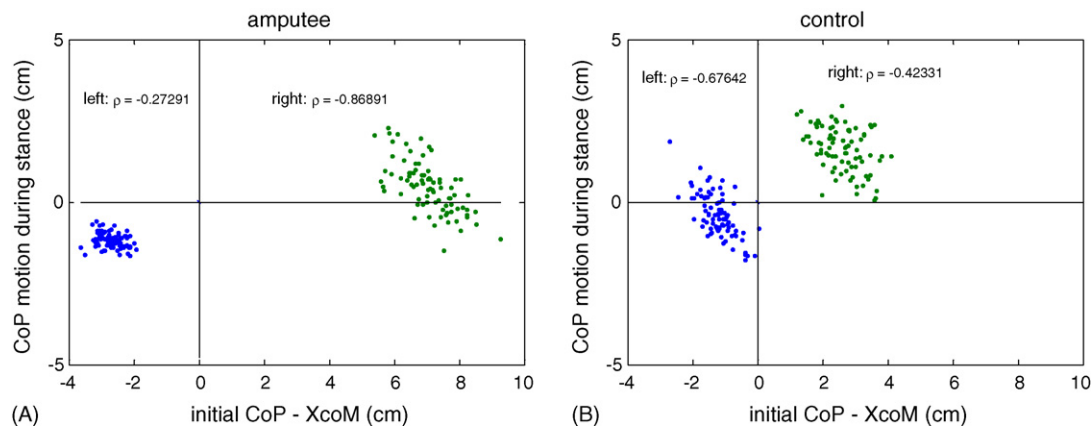


Fig. 6. CoP motion, final minus initial position of the CoP during single stance, as a function of initial distance CoP–XcoM. The control subject and the normal leg in the amputee show a negative correlation, which shows that CoP position is changed during stance in the ‘good’ direction. Correlation coefficients of all subjects are in Table 3.

Table 3
Correlation coefficients between initial CoP–XcoM distance and CoP motion

Subjects	Amputees		Controls	
	P	NP	L	R
1;7	−0.681	−0.789	−0.846	−0.740
2;8	−0.158	−0.895	−0.752	−0.841
3;9	−0.566	−0.682	−0.375	−0.324
4;10	−0.273	−0.869	−0.676	−0.704
5;11	−0.445	−0.844	−0.804	−0.622
6;12	−0.540	−0.387	−0.713	−0.742

4. Discussion

Several of the presented observations confirm earlier findings. The temporal asymmetry between prosthetic and non-prosthetic legs in amputees has been described earlier [19] and it is common knowledge that subjects with a

compromised balance walk with a wider step. According to the traditional idea of stability, it might be concluded that CoM stays within a safe margin of some 3–6 cm from the ‘base of support’, consisting of the left and right CoP positions (Fig. 1). The XcoM concept implies that this view is too optimistic: b_{\min} , the minimum distance between XcoM and average CoP, can in healthy subjects be less than 1 cm, occasionally only 2–3 mm (Table 2 and Fig. 4), comparable to values in standing on one leg.

4.1. Major balance strategies in walking

Not unexpectedly, the stepping strategy turned out to be the most important strategy for lateral balance: when taking a step, the foot has to be positioned within 1–3 cm lateral to the current XcoM (Figs. 1, 3 and 4). Our results in Figs. 5 and 6 and Tables 3 and 4, suggest that the lateral ankle strategy [20] also plays a role. Stepping is a matter of feed forward

Table 4
Standard deviation of initial and average CoP position (cm)

Subjects	Amputees				Controls			
	P		NP		L		R	
	Initial b (cm)	Avg. b (cm)	Initial b (cm)	Avg. b (cm)	Initial b (cm)	Avg. b (cm)	Initial b (cm)	Avg. b (cm)
1; 7	0.53	0.39	0.58	0.25	0.70	0.52	0.64	0.48
2; 8	0.43	0.42	1.05	0.29	0.57	0.39	0.80	0.41
3; 9	0.49	0.36	0.39	0.28	0.46	0.40	0.51	0.39
4; 10	0.37	0.33	0.77	0.29	0.50	0.35	0.65	0.30
5; 11	0.52	0.43	0.46	0.25	0.49	0.28	0.45	0.26
6; 12	0.54	0.35	0.44	0.30	0.41	0.28	0.52	0.27
Mean	0.48	0.38	0.62	0.28	0.52	0.37	0.59	0.35

Standard deviation of the distance between XcoM and CoP, b , at footfall (initial) and averaged over stance (avg.). When the average b is strongly reduced with respect to the initial value, this means that the lateral ankle strategy was effective in reducing b . This is especially seen in the non-prosthetic leg of amputees.

control: the final foot position has to be planned beforehand. During the execution there is limited opportunity to correct the lateral positioning. In a pure stepping strategy, a less correct step can only be corrected in subsequent steps. An ankle strategy can provide minor corrections after the foot has been placed on the basis of feedback. It seems therefore that in normal walking stepping provides gross control and the lateral ankle strategy a fine tuning. The range of the quantity ($u_{\text{initial}} - u_{\text{ms}}$) gives an idea of the extent of the corrections attainable by the ankle strategy. It amounted 0.7–3 cm in control subjects and in amputees 1.7–4.4 cm in the normal leg and 1–2 cm in the prosthetic leg.

From the results of Table 4, it can be seen that initial foot placement is not compromised in amputee walkers, for either leg. The prosthetic leg, however, misses the possibilities of active lateral ankle movement, so that the inaccuracy in average foot placement cannot be corrected in the prosthetic leg. For amputees it is therefore safer to use a wider margin for the XcoM–CoP distance.

The relation between stepwidth and b_{min} can be found from the inverted pendulum model (see Appendix). It is found that:

$$w_{L1-R1} = u_{R1} - u_{L1} = -b_{L1}e^{\omega_0 T_{L1}} + b_{R1} \quad (2)$$

typical values for ω_0 and T are 3 rad/s and 0.55 s, respectively, giving $e^{\omega_0 T} \approx 5$. The width between steps L1 and R1, executed at T_{R1} , can thus be predicted from the previous b_{L1} times a factor 5 plus the actual b_{R1} (see Fig. 7). Such a prediction is in some way necessary, in view of the feed forward control of stepping. A practical problem is that any inaccuracy in b , thus in foot placement, is amplified by a factor of about 5 at the time of the next step.

Eq. (2) gives a number of relevant insights. It shows that stride width w is closely related to the XcoM–CoP margin b_{min} : the wider the margin the wider the step. When foot positioning is less accurate, as in amputees, it is safer to place them further outward, in order to remain stable. This has the disadvantage that the step becomes wider. Another method to increase stability is to decrease T , that is to take faster steps.

Not all steps are equally wide (see Figs. 2 and 3), but in order to walk in a straight path the left–right steps should on

average be equally wide as the right–left steps. If one of the legs functions less precise, with a larger b_{min} , this can be compensated by making the effective stance time T shorter on that side. In amputees we see therefore at the prosthetic side a bigger b_{min} combined with a shorter single stance (see Table 2). This effect may also explain the temporal asymmetry in other one-sided afflictions. Such an asymmetry is thus not a defect, but a sensible adaptation to the one-sided impairment. According to this view, physical therapy should thus not aim at improving temporal symmetry in all patients with asymmetric impairments. The temporal asymmetry in our amputee walkers supports this. All of them were able walkers and all had many years of experience in walking with a prosthesis. The negative outcome of the Stroop test just confirms that walking did not require attention for them.

The formula (2) has been used to predict the actually measured stride width from the previous and the current b_L , b_R , and T . The choice of T needed some consideration. In principle, it should amount to the single stance period. In the

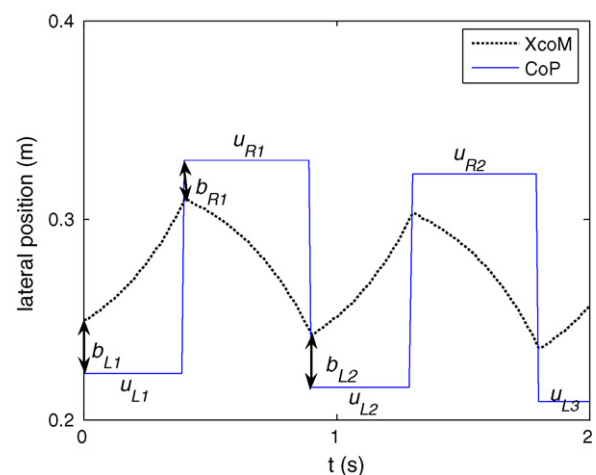


Fig. 7. Illustration of Eqs. (A.6) and (A.7). During the first left step, with the CoP at $u = u_{L1}$, the XcoM moves with an exponential course to the right, Eq. (A.7). The right foot is placed with an additional margin b_{R1} lateral to the XcoM at the instant of foot contact. This determines the CoP position of the right foot, u_{R1} . After this the process repeats itself to the left, etc. As a consequence, the width of this stride, $w_1 = u_{R1} - u_{L1}$, is determined by the two margins b_{L1} and b_{R1} , Eq. (2).

Table 5
Model predictions

Subjectst	Mean error (cm)				r.m.s. error (cm)			
	Amputees		Controls		Amputees		Controls	
	NP	P	L	R	NP	P	L	R
1; 7	0.743	−0.478	0.234	0.033	1.424	0.563	0.673	0.410
2; 8	−1.086	0.979	0.394	0.614	1.243	0.996	0.715	0.434
3; 9	1.442	−0.440	0.168	0.289	0.438	0.745	0.402	0.277
4; 10	−0.362	1.586	0.123	1.158	0.796	0.543	0.482	0.404
5; 11	0.920	1.029	0.508	0.687	0.614	0.590	0.327	0.273
6; 12	1.618	−0.101	0.123	0.199	0.951	0.644	0.354	0.323
Mean	0.546	0.429	0.258	0.497	0.973	0.698	0.515	0.360

Mean error and standard deviation of error in the prediction of step width by the inverted pendulum model (7). The root-mean-square (r.m.s.) error was higher for the prosthetic leg vs. the normal leg in amputees ($p < 1\%$), and higher in amputees than in controls ($p < 1\%$).

simple model as presented it is assumed that the CoP switches instantaneously from left to right, i.e. that the double stance period is zero. As to be expected, inserting the single stance (=contralateral swing) duration for T resulted in too small estimates of step width. It was better to use for T values equal to single stance $+0.5 \times$ double stance. Results of the prediction can be seen in Table 5. In the control subjects the root-mean-square (r.m.s.) error was of the order of 3–7 mm, to be compared to stride widths of 86 mm. Next to this, there was a small mean error. We might even have reduced the mean error to an average zero, by some fiddling of the T , but this was not attempted. In any case, the predictions by the simple inverted pendulum model can be considered very good for normal subjects.

In the amputee subjects model predictions were less good, a higher r.m.s. error and greater differences of the mean error between subjects. This means that the strategies related to the inverted pendulum model, stepping and ankle strategy, are not sufficient to completely explain amputee gait. Additional actions must play a role as well. Possible candidates are movements of the trunk and holding the hands. The amputee subjects felt not sufficiently comfortable with treadmill walking to walk without touching at least one handrail. Trunk movements were also evident: the video recordings showed in all amputee subjects to some degree specific trunk movements, with the upper trunk moving in opposite phase with the pelvis, similar to the original description by Trendelenburg [21], which were not observable in any of the control subjects. Our impression is that this trunk motion is primarily a form of adaptation to walking with a prosthesis, and less a strategy for balance control, but this is a point to be investigated further. Nevertheless, this trunk rotation will certainly influence IP-model predictions, as the IP-model does not take such rotations into account.

5. Conclusions

- (1) In order to walk stable and in a straight path, the foot (CoP) has to be placed with an accuracy of a few millimetres. In normal subjects this is accomplished in

gross control by a stepping strategy, with a fine tuning by the lateral ankle strategy.

- (2) The temporal asymmetry in amputee walking can be considered as a sensible adaptation to the impairment due to the prosthesis. This suggests that physical therapy should not aim at improving temporal symmetry in amputees or in patients with asymmetric impairments in all cases.

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Appendix A

A.1. Prediction of step width from the inverted pendulum model

The relation between step width and XcoM–CoP margin can be understood from the ‘inverted pendulum model’ for walking. In this model the human body is modelled as a single mass at the CoM, balancing on a stick of length h , the lower end of which is positioned at the CoP. A major assumption of this model is that limbs or trunk do not rotate with respect to the whole-body CoM [22]. The inverted pendulum model can be formulated by the second order differential equation:

$$\ddot{z} = (z - u) \frac{h}{g} \quad (\text{A.1})$$

For constant u and initial values x_0 and v_0 Eq. (A.1) can be solved to give (see Nomenclature for symbols).

$$z(t) = u + (z_0 - u) \cosh(\omega_0 t) + \frac{v_0}{\omega_0} \sinh(\omega_0 t) \quad (\text{A.2})$$

The z -velocity is thus:

$$\dot{z}(t) = (z_0 - u) \omega_0 \sinh(\omega_0 t) + v_0 \cosh(\omega_0 t) \quad (\text{A.3})$$

For XcoM position $\zeta(t)$ then follows:

$$\zeta(t) = z(t) + \frac{\dot{z}(t)}{\omega_0} = (\zeta - u)e^{\omega_0 t} + u = -b_{\min}e^{\omega_0 t} + u \quad (\text{A.4})$$

In walking the time course of $u(t)$ can be approximated as an alternation of steps with u constant during the step, left $u = u_{L1}, u_{L2}, \dots$ and right $u = u_{R1}, u_{R2}, \dots$ with durations $T_{L1}, T_{R1}, T_{L2}, T_{R2}, \dots$ (see Fig. 7). For the second step it is seen that:

$$u_{R1} = \zeta(T_{L1}) + b_{R1} \quad (\text{A.5})$$

Together with (A.4) this gives:

$$u_{R1} = -b_{L1}e^{\omega_0 T_{L1}} + u_{L1} + b_{R1} \quad (\text{A.6})$$

For the width of the first step thus holds:

$$w_{L1-R1} = u_{R1} - u_{L1} = -b_{L1}e^{\omega_0 T_{L1}} + b_{R1} \quad (\text{A.7})$$

and so on for the following steps.

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